

PNAS

^cDefense and Veterans Brain Injury Center, Walter Reed Army Medical Center, Washington, DC 20307

Contributed by John D. Joannopoulos, October 8, 2010 (sent for review September 13, 2010)

computational models | personal protection equipment

¹To whom correspondence may be addressed. E-mail: ioannop@mit.edu or rapa@mit.edu.

Report Documentation Page				Form Approved OMB No. 0704-0188	
Public reporting burden for the collection of information is estimated to average 1 hour per response, including the time for reviewing instructions, searching existing data sources, gathering and maintaining the data needed, and completing and reviewing the collection of information. Send comments regarding this burden estimate or any other aspect of this collection of information, including suggestions for reducing this burden, to Washington Headquarters Services, Directorate for Information Operations and Reports, 1215 Jefferson Davis Highway, Suite 1204, Arlington VA 22202-4302. Respondents should be aware that notwithstanding any other provision of law, no person shall be subject to a penalty for failing to comply with a collection of information if it does not display a currently valid OMB control number.					
1. REPORT DATE SEP 2010		2. REPORT TYPE		3. DATES COVERED 00-00-2010 to 00-00-2010	
4. TITLE AND SUBTITLE In silico investigation of intracranial blast mitigation with relevance to military traumatic brain injury				5a. CONTRACT NUMBER	
				5b. GRANT NUMBER	
				5c. PROGRAM ELEMENT NUMBER	
6. AUTHOR(S)				5d. PROJECT NUMBER	
				5e. TASK NUMBER	
				5f. WORK UNIT NUMBER	
7. PERFORMING ORGANIZATION NAME(S) AND ADDRESS(ES) Walter Reed Army Medical Center, Defense and Veterans Brain Injury Center, Washington, DC, 20307				8. PERFORMING ORGANIZATION REPORT NUMBER	
9. SPONSORING/MONITORING AGENCY NAME(S) AND ADDRESS(ES)				10. SPONSOR/MONITOR'S ACRONYM(S)	
				11. SPONSOR/MONITOR'S REPORT NUMBER(S)	
12. DISTRIBUTION/AVAILABILITY STATEMENT Approved for public release; distribution unlimited					
13. SUPPLEMENTARY NOTES					
14. ABSTRACT					
15. SUBJECT TERMS					
16. SECURITY CLASSIFICATION OF:			17. LIMITATION OF ABSTRACT Same as Report (SAR)	18. NUMBER OF PAGES 6	19a. NAME OF RESPONSIBLE PERSON
a. REPORT unclassified	b. ABSTRACT unclassified	c. THIS PAGE unclassified			

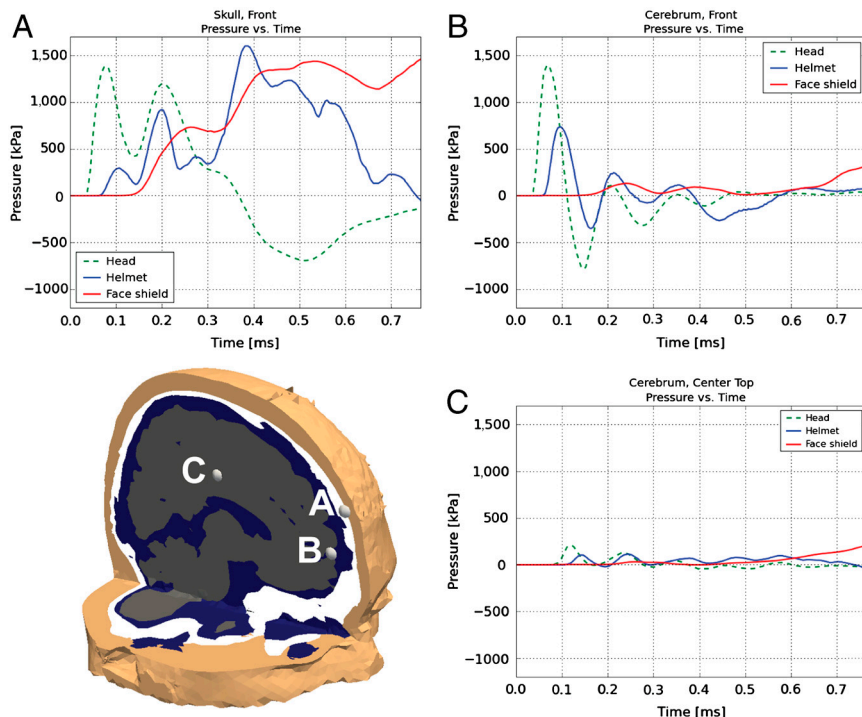


Fig. 2. Pressure profiles from points in the cerebrum and skull for the head, helmet, and face shield simulations. The scale is from $-1,200$ to $1,700$ kPa.

in the intracranial cavity have largely dissipated, with the largest remaining stresses located in the skull.

In the helmet simulation (center column of Fig. 1), it can be seen that the presence of the helmet slightly delays the arrival of the blast wave but does not impede direct transmission of stress waves into the intracranial cavity because it does not protect the face. The snapshot at $t = 0.231$ ms shows that the negative phase of the blast wave still causes underpressure behind the eyes, but the presence of the helmet avoids the underpressure in the CSF observed in the unhelmeted head simulation. By contrast, slightly higher pressures are transmitted to the skull via the foam padding elements, which act as load concentration points for the impulse transferred from the blast wave to the surface of the helmet. Because it does not cover the face, the helmet does not significantly contribute to mitigating the stress waves transmitted to the brain tissue. Although it does protect the top part of the head from direct exposure to the blast, the advantage is minimal because this is not a major pathway of load transmission into the intracranial cavity. Conversely, these results show that the existing ACH does not enhance blast effects on brain tissue as has been suggested recently with more simplistic models (19).

The column on the right of Fig. 1 shows snapshots of the simulation including face protection. It can be seen that the immediate effect of the face shield is to impede direct transmission of stress waves to the face, both for the positive and negative phases, $t = 0.059, 0.232$ ms. Initially, the main load transmission pathway to the head is via the foam padding, $t = 0.232$ ms. This

causes a delay in the transmission of stress waves into the head. The magnitude of the transmitted stress is also limited by the crushing stress of the foam, which can be made significantly lower than the reflected blast overpressure, thus affording a passive (soft catch) mitigation mechanism, as discussed recently in ref. 25. By contrast, the total impulse transmitted via the foam padding increases by the extra amount transmitted to the face shield surface. The late increase of the pressure on and inside the head can be attributed to the external shock wave wrapping around the back of the face shield surface and into the interstitial volume between the head and the shield. This points to the need to improve this particular conceptual design.

To develop a more quantitative understanding of the simulation results, pressure histories at three points within the skull and cerebrum were extracted and compared for all three simulations, Fig. 2. It can be observed that the helmet alone only slightly delays and reduces the magnitude of pressure peaks, whereas the mitigating effect of the helmet–face shield combination is much more pronounced. For example, although point B in the front of the cerebrum experiences an initial pressure peak of $1,392$ kPa at 0.067 ms in the head simulation, in the helmet simulation the peak is delayed by 0.03 ms and reduced to 734 kPa. In the face shield simulation, the same peak is delayed by an additional 0.141 ms and the magnitude is reduced to 132 kPa, a tenth of the magnitude in the head simulation. Although the specific impulse (area underneath the pressure vs. time curve) in the skull (point A) is greater in the helmet and face shield simulations than

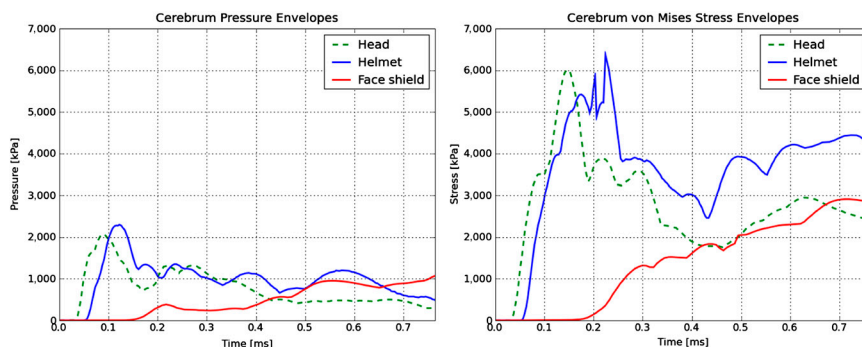


Fig. 3. Pressure and von Mises stress envelopes for the cerebrum. The scale is from 0 to $7,000$ kPa.

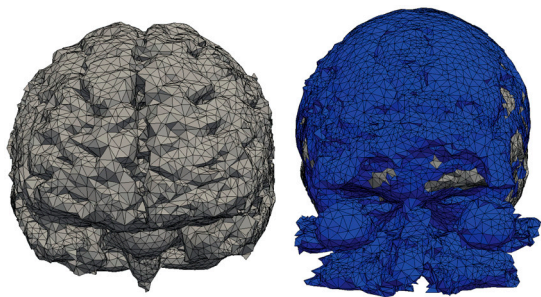


Fig. 4. Computational mesh of the consolidated cerebrum and CSF structures.

in the unprotected case, as expected from the additional impulse transmission surface, the delay and reduction in magnitude of the early blast response is significant. Fig. 2 also suggests that the highest pressures occur in the skull (point A), which, due to its stiffness, provides some natural protection to the brain tissue. Inside the brain, the maximum stresses are observed in the anterior region, where the transmitted stress waves initially enter the head for the front blast case analyzed.

Fig. 3 shows pressure and von Mises stress envelopes (plots of the highest value at each time step) in the cerebrum for the three simulations. These figures further confirm that the helmet slightly delays and does not significantly increase or mitigate stresses transmitted to the brain tissue, whereas a face-protecting device does.

Discussion

A simulation-based investigation of possible strategies to mitigate the effects of blasts on the human head was conducted. A first simulation was run using an advanced model of the human head without protection exposed to a frontal blast wave of intensity comparable to threshold values of blast lung injury (23, 26). The results suggest that the main wave transmission pathways are the soft tissues in direct contact with the incident blast wave. This simulation also showed that cavitation, if possible, is more likely to be associated with endogenous wave reflections than with the negative phase of the incident blast wave. Tissue cavitation has been proposed as a potential physical mechanism conducive to brain injury (27–29). The second simulation was intended to evaluate the blast protection properties provided by the ACH. The results suggest that the ACH provides no significant mitigation of blast effects on brain tissue. However, no significant deleterious wave-focusing effects were observed either. A third simulation included a conceptual face shield rigidly attached to the helmet shell. It was found that the presence of this device contributed significantly to reducing the magnitude of the stresses propagated inside the brain. However, the particular preliminary design of the face shield adopted needs significant optimization.

The study was limited to a single set of material and blast characteristics (frontal incidence, fixed explosive mass, type, and standoff), which was sufficient to establish theoretical evidence that covering the exposed head surfaces will likely contribute

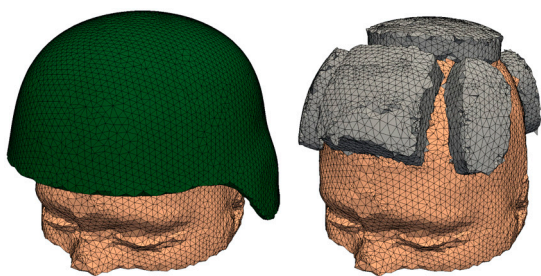


Fig. 5. Head model with ACH shell and padding.



Fig. 6. Head model with ACH and conceptual face shield: (A) geometry of the face shield and (B) combined sagittal and axial cut showing a detail of the full mesh.

to mitigating blast-induced mild TBI. The conclusions are based on the trends and differences observed among the three simulations, which clearly show the effect of protection equipment. The computational framework has been validated in nonbiological systems, e.g. (25, 30). Current experimental validation efforts involve shock-tube tests on in vivo instrumented porcine specimens, in vitro laboratory-scale blast tests, as well as free-field blast tests. The validation process involves a comparison of the simulation results with the recorded experimental data consisting of pressure histories at sensor locations outside and inside the head for different blast standoff distances. Extensions to the model will include the neck and torso, which have been suggested as a possible indirect pathway for blast-induced brain injury (31). Optimization of the face shield to achieve specific mitigation targets will be explored via parametric studies considering more general blast conditions (incidence angle and intensity), face shield geometries, and material properties. Improvements may include increasing the stiffness and extending the face shield in the posterior lower region of the head to reduce wave diffraction around the tip.

Materials and Methods

Computational Framework for Simulating Blast–Head–Helmet Interactions. The simulations were conducted using an extension of the Virtual Test Facility (VTF) (30, 32) to simulate blast wave–structure interactions. The VTF is a suite of integrated computational fluid and solid mechanics solvers for the fully coupled analysis of the response of solids to blast and detonation-wave loading on massively parallel computing platforms. Constitutive models to describe the response of various tissues and biological structures have been added to the solid mechanics solver. In addition, the capability to simulate blast waves in air of arbitrary intensity has been integrated in the code to initialize the simulations. The characteristics of the blast wave are specified by the type, mass, and spatial location of the explosive.

The DVBIC/MIT Full Head Model and Extensions Including Protective Equipment.

The three-dimensional computational model of the human head developed by DVBIC and MIT (17) was adapted to this study's requirements and used in the simulations. It consists of a biofidelic representation of the human head including the following 11 distinct structures: skin and fat, muscle, skull, air sinus, eyes, CSF, gray matter, white matter, ventricles, venous sinus, and glia. The model was obtained by reconstruction from high-resolution T1 MR images and bone-windowed computed tomography images of a human head via registration, segmentation, and posterior computational mesh generation. For the purpose of the current study, the 11 structures were consolidated into 4: cerebrum, skull, CSF, and soft tissue.

A computer-aided design model of the actual ACH shell and pad geometries provided by Natick Soldier Research Development and Engineering Center was combined with the head model into a single computational model of the head/helmet system. A simple face shield geometry was then created such that the top edge of the face shield was coincident with the

Table 1. Material properties for skull

Material	Density, kg/m ³	K, Pa	G, Pa	κ, Pa·s	μ, Pa·s	C ₀ , m/s	s
Skull	1,412	3.86 × 10 ⁹	2.665 × 10 ⁹	0.0	0.0	1,850.0	0.94

Table 2. Material properties for CSF, cerebrum, and skin/fat

Material	Density, kg/m ³	K, Pa	G, Pa	κ , Pa·s	μ , Pa·s	Γ_0
CSF	1,040	2.19×10^9	4.38×10^2	1.0×10^3	0.0	6.15
Cerebrum	1,040	2.19×10^9	2.253×10^4	1.0×10^3	0.0	6.15
Skin/fat	1,100	3.479×10^7	5.88×10^6	1.59×10^1	0.0	6.15

front edge of the helmet. The geometry of the face shield was designed to smoothly project from the surface of the helmet shell.

The main characteristics of the head, helmet, and face shield computational models are illustrated in Figs. 4 (cerebrum and CSF), 5 (head surface, ACH, and foam pads), and 6 (face shield and interior view of fully assembled model).

All finite element meshes were generated using the Octree algorithm in Ansys. Bad quality tetrahedra, which were commonly obtained, were eliminated using the HealMesh* mesh optimization library. The resulting mesh of the complete head model with helmet and face shield consists of 443,452 tetrahedra with quadratic interpolation (10 nodes). In the simulations, 30 processors were used for the solid solver and 10 for the fluid solver. The fluid grid used two levels of subdivision with an equivalent resolution of $1,200 \times 1,000 \times 580$ grid points.

Material Models and Properties. The constitutive models adopted in the simulations were selected to accurately describe the propagation of stress waves inside the head and protective structures transmitted from the air blast wave. To this end, suitable equations of state describing the volumetric (pressure) response were adopted for each tissue or material type. In addition, the deviatoric recoverable response was described using a nonlinear elastic model. Finally, dissipative effects in the brain tissue were considered through a linear viscosity model. The advantage of this simplified constitutive modeling approach is that it requires few physical material or tissue parameters, which can be quantified with some certainty.

For the skull, we adopted the Hugoniot equation of state (33), which is widely used to describe the shock response of many solid materials:

$$p = \frac{\rho_0 C_0^2 (1 - J)}{[1 - s(1 - J)]^2}. \quad [1]$$

In this expression, p is the pressure, J is the local volume change given by the determinant of the deformation gradient tensor, and C_0 and s are material parameters. For the remaining head structures, we adopted the Tait equation of state, which is commonly used to model fluids subject to large pressure variations (34):

$$p = B[J^{-(\Gamma_0+1)} - 1]. \quad [2]$$

- Warden D (2006) Military TBI during the Iraq and Afghanistan wars. *J Head Trauma Rehab* 21:398–402.
- Tanielian T (2008) *Invisible Wounds of War: Psychological and Cognitive Injuries, Their Consequences, and Services to Assist Recovery* (RAND Corporation, Center for Military Health Policy Research, Santa Monica, CA).
- Moore DF, et al. (2008) Blast physics and central nervous system injury. *Future Neurology* 3:243–250.
- Moore DF, Radovitzky R, Jerusalem A, Nyein M, Jaffee M (2008) Military traumatic brain injury (TBI), blast and impact injury. *Ann Neurol* 64:530.
- Hoge CW, et al. (2008) Mild traumatic brain injury in U.S. soldiers returning from Iraq. *New Engl J Med* 358:453–463.
- Terrio H, et al. (2009) Traumatic brain injury screening: Preliminary findings in a U.S. army brigade combat team. *J Head Trauma Rehab* 24:14–23.
- Defense and Veterans Brain Injury Center (DVBIC) (2010) TBI numbers: Department of Defense numbers for traumatic brain injury. <http://www.dvbic.org/TBI-Numbers.aspx>.
- Langlois J, Rutland-Brown W, Thomas K (2006) *Traumatic Brain Injury in the United States: Emergency Department Visits, Hospitalizations, and Deaths* (Centers for Disease Control and Prevention, Natl Center for Injury Protection, Atlanta, GA).
- Warden D, et al. (2005) War neurotrauma: The defense and veterans brain injury center (DVBIC) experience at Walter Reed Army Medical Center (WRAMC). *J Neurotrauma* 22:1178.
- Ling G, Bandak F, Armonda R, Grant G, Ecklund J (2009) Explosive blast neurotrauma. *J Neurotrauma* 28:815–825.
- Murray C, et al. (2005) Spectrum of care provided at an Echelon II medical unit during Operation Iraqi Freedom. *Mil Med* 170:516.
- Gawande A (2004) Casualties of war—military care for the wounded from Iraq and Afghanistan. *New Engl J Med* 351:2471–2475.
- Rosen J, et al. (2008) *IED Task Force Medical Panel* (Defense Science Board, Irvine, CA) technical report.

Table 3. Material properties for helmet, face shield, and padding used in simulations

Material	Density, kg/m ³	E, Pa	ν
Helmet/face shield	1,440	1.24×10^9	0.36
Padding	136	8×10^6	0.2

In this expression, $B = K/(\Gamma_0 + 1)$ and Γ_0 are material parameters. The Tait equation provides a reasonable representation of the volumetric response of soft tissues embedded in a fluid medium.

The deviatoric elastic response $\sigma^{e,dev}$ was computed using the neo-Hookean model extended to the compressible range, in which the strain energy density is given by

$$W(C) = \frac{\lambda}{2} \log^2 J - \mu \log J + \frac{\mu}{2} (I_1 - 3), \quad [3]$$

where μ and λ are Lamé constants and I_1 is the first invariant of the right Cauchy–Green deformation tensor.

To complete the constitutive description of each different tissue or material structure, a linear viscosity model was added to both the deviatoric and the volumetric response, furnishing a final expression for the Cauchy stress components:

$$\sigma_{ij} = \sigma_{ij}^{e,vol} + \sigma_{ij}^{e,dev} + 2\mu_v d_{ij}^{dev} + \kappa d_{ii} \delta_{ij}, \quad [4]$$

where d_{ij} are the components of the rate of deformation tensor and μ_v , κ , are, respectively, the deviatoric and volumetric viscosity parameters.

The model parameters used for the head components were obtained from literature data (17), Tables 1 and 2.

Given the intensity of the blast waves under consideration, the response of the engineering materials used in the protective structures was expected to stay well in the elastic regime. Consequently, the use of a simple neo-Hookean elastic model for these materials was justified for this application. Standard properties for the Kevlar shell and foam were respectively used for the helmet/face shield and padding, Table 3.

ACKNOWLEDGMENTS. This work was supported by financial aid from the Joint Improvised Explosive Device Defeat Organization through the Army Research Office. Partial support from the MIT Institute for Soldier Nanotechnologies is also gratefully acknowledged.

*HealMesh © Parasim Inc.

- Willinger R, Baumgartner D (2003) Human head tolerance limits to specific injury mechanisms. *Int J Crashworthiness* 8:605–617.
- Zhang L, Yang K, King A (2004) A proposed injury threshold for mild traumatic brain injury. *J Biomech Eng—T ASME* 126:226–236.
- Horgan T, Gilchrist M (2004) Influence of FE model variability in predicting brain motion and intracranial pressure changes in head impact simulations. *Int J Crashworthiness* 9:401–418.
- Moore DF, et al. (2008) Computational biology, modeling of primary blast effect on the central nervous system. *NeuroImage* 47:T10–T20.
- Taylor P, Ford C (2009) Simulation of blast-induced early-time intracranial wave physics leading to traumatic brain injury. *J Biomech Eng* 131:061007.
- Moss W, King M, Blackman E (2009) Skull flexure from blast waves: A mechanism for brain injury with implications for helmet designs. *J Acoust Soc Am* 125:2650–2667.
- Jordan P, Socrate S, Zickler TE, Howe RD (2008) *A Nonrigid Image Registration Framework for Identification of Tissue Mechanical Parameters, Medical Image Computing and Computer-Assisted Intervention* Lecture Notes in Computer Science, eds D Metaxas, L Axel, G Fichtinger, and G Székely (Springer, Berlin), Vol 4190, pp 930–938.
- Pervin F, Chen W (2009) Dynamic mechanical response of bovine gray matter and white matter brain tissues under compression. *J Biomech* 42:731–735.
- Alley MD, Son SF (2009) *Blast Loading Experiments of Developed Surrogate Models for TBI Scenarios*, (Am Physical Society, Nashville, TN), Vol C2.00001.
- Bowen I, Fletcher E, Richmond D, Hirsch F, White C (1968) Biophysical mechanisms and scaling procedures applicable in assessing responses of the thorax energized by air-blast overpressures or by nonpenetrating missiles. *Ann NY Acad Sci* 152:122–146.
- Malvern L (1969) *Introduction to the Mechanics of a Continuous Medium* (Prentice-Hall, Englewood Cliffs, NJ).
- Wadley HNG, et al. (2010) An active concept for limiting injuries caused by air blasts. *Int J Impact Eng* 37:317–323.
- Bass C, Rafaels K, Salzar R (2008) Pulmonary injury risk assessment for short-duration blasts. *Journal of Trauma* 65:604–615.

